35th Annual International Conference of the IEEE EMBS
Osaka, Japan, 3 - 7 July, 2013

Piezoelectric and Electromagnetic Respiratory Effort Energy Harvesters*

Ehsaneh Shahhaidar, *Student Member, IEEE,* Bryson Padasdao, *Student Member, IEEE*, R. Romine,
C. Stickley, Olga Boric-Lubecke, *Senior Member, IEEE*

***Abstract*—The movements of the torso due to normal breathing could be harvested as an alternative, and renewable power source for an ultra-low power electronic device. The same output signal could also be recorded as a physiological signal containing information about breathing, thus enabling self-powered wearable biosensors/harvesters. In this paper, the selection criteria for such a biosensor, optimization procedure, trade-offs, and challenges as a sensor and harvester are presented. The empirical data obtained from testing different modules on a mechanical torso and a human subject demonstrated that an electromagnetic generator could be used as an unobtrusive self-powered medical sensor by harvesting more power, offering reasonable amount of output voltage for rectification purposes, and detecting respiratory effort.**

I. INTRODUCTION

Dependency of the ubiquitous computing and wireless sensor networks on batteries presents a concern when there is need to recharge or change the batteries. This issue becomes critical when it comes to remote health monitoring. The limited battery storage ability impedes the growth of such applications. Human energy as one form of renewable energy opens the way for the development of technologies that can harvest this energy for powering low power personal electronic devices.

Although human energy harvesting to power a wearable computer was proposed about a decade ago [1], most promising recent efforts have been focused on kinetic energy harvesting during walking [2-7]. However, there is another type of human energy, simply the movements of the torso due to breathing, which could be harvested as an alternative power source for ultra-low power wearable biosensors.

At the same time, movements of the chest wall due to breathing could reveal critical medical parameters that can predict catastrophic events. It has been reported that abnormal physiological observations begin 6-8 hours prior to cardiopulmonary arrest [8, 9]. In these studies abnormal respiration is often identified as a sensitive clinical predictor for serious adverse events [10, 11]. Respiratory failure is the most common cause of ICU admission from general hospital wards. Yet unfortunately, observation of respiration is infrequently performed, even when the primary problem is a respiratory condition [12, 13]. Direct measurements of respiratory effort, and changes in lung volume are performed by esophageal manometers and spirometers respectively, which are obtrusive and are not used routinely. An unobtrusive surrogate measure of respiratory effort can be obtained by measuring changes in chest and/or abdominal volume, like plethysmography [14].

We have combined the idea of harvesting the energy originated from movements of the torso during breathing and recording the volume change or simply circumferential change of the thoracic area (including chest and abdomen), to produce a self-powered biosensor by reusing the hardware components to perform both harvesting and sensing functions. Potential applications of this kind of zero-net energy wearable biosensor will be remote health monitoring. Introduction of an autonomous respiratory monitor could increase detectability of deteriorating patients, increase early therapeutic intervention, and potentially decrease serious adverse events as ICU admission and cardiopulmonary arrest. However, a desirable human energy harvester would have the least interference with the normal daily activities, i.e. it parasitically harvests energy and it is unobtrusive.

One way to harvest power parasitically from breathing is using a chest belt including power scavenging modules [15-17]. Investigations show that the available harvestable power from circumferential change of the torso due to normal breathing without putting an onerous load on the user is on the order of mW [18]. In [15], the theoretical feasibility of using a piezoelectric bending generator to concurrently sense the respiratory effort and harvest energy was proposed. In [16], the idea of using a miniature DC generator as a harvester was simulated with a maximum output voltage of 0.1V and power of 2mW. An electromagnetic servomotor with a high gear ratio has been also been used [17] for the purpose of sensing and harvesting simultaneously, being able to produce higher voltage (0.7V) and power (15mW) with the trade-off of higher required force from the chest to move the motor armature.

This paper outlines factors that have to be considered when designing a self-powered biosensor, and the procedure of selecting the sensing/harvesting materials. The proposed harvesting modules are tested on a mechanical torso, and the most efficient one is also tested on a human subject.

---

* This work was supported in part by the National Science Foundation under grant ECCS-0926076, the University of Hawaii Renewable Energy and Island Sustainability (REIS) Program, and by Department of Energy grant, DE-OE0000394.

Ehsaneh Shahhaidar is a PhD student at the Electrical Engineering Department, University of Hawaii at Manoa, Honolulu 96822 USA (e-mail: ehsaneh@hawaii.edu).

Bryson Padasdao is a PhD student at the Electrical Engineering Department, University of Hawaii at Manoa, Honolulu 96822 USA (e-mail: brysonep@hawaii.edu).

Rebecca Romine is a PhD student at the Kinesiology and Rehabilitation Science Department, University of Hawaii at Manoa, Honolulu 96822 USA (email: rromine@hawaii.edu).

Cris Stickley is an Assistant Professor at the Kinesiology and Rehabilitation Science Department, University of Hawaii at Manoa, Honolulu 96822 USA (email: cstickle@hawaii.edu).

Olga Boric-Lubecke is a Professor at Electrical Engineering Department, University of Hawaii at Manoa, Honolulu 96822 USA (e-mail: olgabl@hawaii.edu).

Experimental results indicate that sufficient energy can be obtained by our electromagnetic respiratory effort sensor/harvester to power a low power System on Chip, such as TI MSP430 [19], while providing more comfort as a sensor.

II. SENSOR/HARVESTER DESIGN AND IMPROVEMENT

Designing an efficient multipurpose device, such as a concurrent physiological sensor and a harvester, which is also unobtrusive and wearable, requires a multifactor optimization. The efficacy of the device as a sensor depends on the accuracy of detected signals, while as a harvester the efficiency is directly related to the output voltage and power. Another important factor is that the device should be ergonomic, and no extra load and effort should be applied to the body while wearing the device, hence the sensed signal will represent the real physiological activity and it will be widely adopted by people. Thus the design process includes some trade-offs.

A. Piezoelectric Respiration Sensor/Harvester

The most common, inexpensive, and simple method in current use for measuring the changes of chest movements during breathing is to use a piezoelectric sensor. Piezoelectric sensor is a crystal that directly generates a voltage when compressed or stretched [15]. Fig. 1 shows the block diagram of a piezoelectric energy harvesting module. We initially tested two different types of piezoelectric beam benders to make our sensor/harvester module, obtaining information about respiratory effort and harvesting the same signals as a power source. Both sensor/harvester systems were embedded in a belt alongside the chest.

First, Double Quick-Mount PZT-5A4E was tested (Fig.2). The results proved that this type of material works well as a respiratory effort sensor [15, 16].

Then another more comfortable piezoelectric strip, MFC, was tested. MFC, (Micro Fiber Composite) M-8528-P2, is a very flexible, lightweight, planar piezoelectric device consisting of rectangular cross-section PZT fibers embedded in an epoxy matrix and sandwiched between Kapton®-clad copper interdigitated electrodes [20]. MFC was a better and more accurate sensor since it was also able to detect the small movements of the chest due to heart beat (Fig.3). Both piezoelectric devices were offering accurate, comfortable and unobtrusive respiratory effort sensors, although MFC proved to be more comfortable, while offering lower voltage compared to PZT.

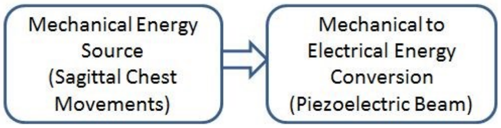

Figure 1. Piezoelectric energy harvesting block diagram

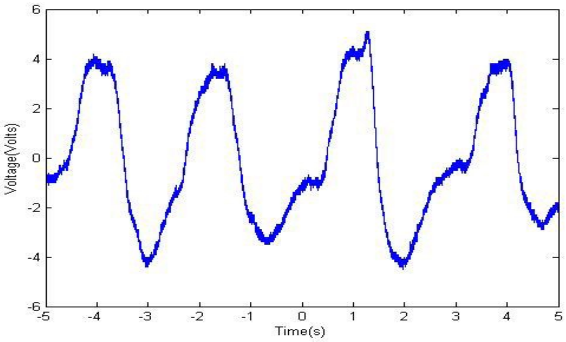

Figure 2. Respiratorion signal detected by PZT-5A

With high load resistances, the output voltages of both tested materials were high enough to be rectified with a normal diode bridge (4.25V, and 1.5V peak for PZT and MFC respectively), but due to low capacitance of both piezo materials (CPZT=232nF, CMFC=172nF), the output current, and thereafter overall harvested energy was low, as indicated in Table I. Since the harvestable power was prohibitively low, more efficient energy scavenging methods were investigated.

B. Improved Electromagnetic Sensor/Harvester

Electromagnetic energy harvesting typically produces electricity with a higher efficiency than piezoelectric methods [5-7, 16, 17]. Linear displacement caused by changes in chest circumference can be converted to electricity by either a linear or rotary electromagnetic generator creating a voltage signal proportional to the chest motions. Fig. 4 depicts the block diagram of a rotational electromagnetic energy harvester. An initial prototype for respiratory effort harvesting was tested in [16] while the output voltage (100mV) was not high enough for rectification purposes. Also, a low load resistance (10ohms) was used to provide high output current, while resulting in high required force for the armature to be turned [17]. Thus the design was improved by considering the 1) unobtrusiveness by minimizing required force from the torso to pull the belt connected to the armature, 2) increased output voltage, and 3) increased harvested power.

Less required mechanical force in a rotational electromagnetic machine means lower torque or a longer coupling arm. Longer coupling arm is not practical because of the size limitations of the device. Lower torque either requires lower gear ratio, or lower current (1), hence higher load resistance:

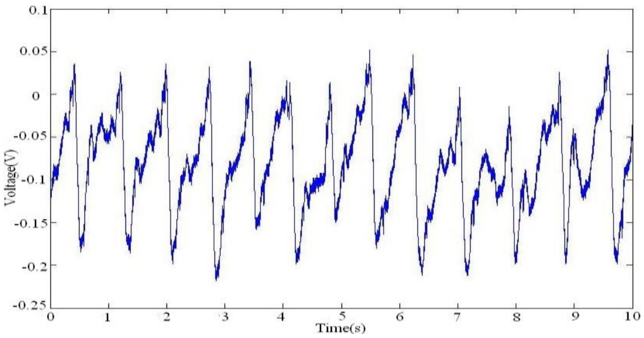

Figure 3. Heart beat signal detected by MFC

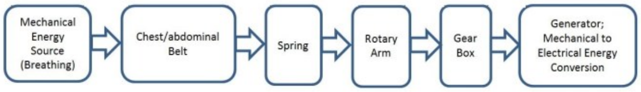

Figure 4. Rotational electromagnetic energy harvesting block diagram

$$\tau = I.k_M, \qquad (1)$$

where k_M is the machine's torque constant [21].

Higher load resistance results in higher output voltage, which is also desirable. However, using a higher load resistance will decrease the delivered power. On the other hand, decreasing the gear ratio lowers the required force but both output voltage and power will be increased; in case of increasing the load resistance the required force will become less, the output voltage will be increased but the electrical output power will be decreased.

Fig. 5 shows the measured required force vs. load resistance, and Fig. 6 the measured output current vs. output voltage for different load resistances during normal breathing, for miniature electromagnetic metal gear head generators with 30:1 and 50:1 gear ratios. The maximum force applied from the chest wall to the skin, i.e. available force, during normal breathing has been reported to be around 0.45N to 3.53N for a healthy subject depending on the direction of the force [18]. Since this amount varies from person to person, and also to be able to pick up almost all respiration movements by the sensor/harvester, the lower limit of the available force was considered. To obtain a higher output voltage and power the 50:1 gear ratio was selected. Higher gear ratio generators were not practical to be used as a low-force-required device. Also, the optimized selected load resistance; based on the required force in order to turn the armature, the output voltage, and the output power was selected to be 160 Ω.

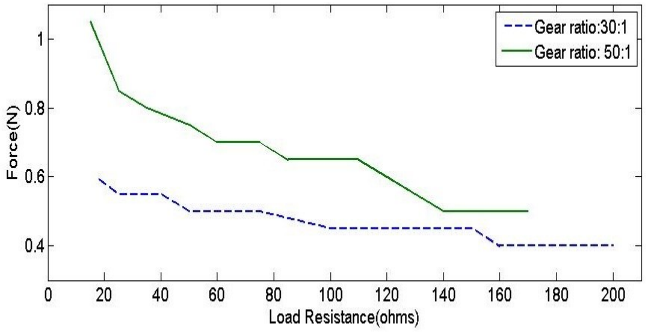

Figure 5. The measured required force vs. load resistance for the same type of electromagnetic generator with 30:1, and 50:1 gear ratios

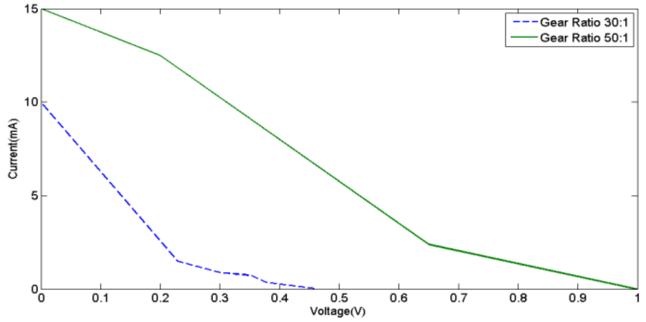

Figure 6. Measured output current vs. output voltage during normal respiration based on the load resistance change for the same type of electrmagnetic generator with 30:1, and 50:1 gear ratios

III. EXPERIMENTAL RESULTS

To have an accurate comparison, all three modules, the strips of PZT-5A, and MFC, and a Pololu Micro Metal Gearmotor (G=50:1) were attached to a mechanical torso while an output load of 160 Ω was attached to each of them. The generator was connected to an adjustable wire strap with a small spring at one end of the strap to help the retracting process during exhalation and contraction of the torso. The mechanical torso was programmed to move at 0.5cm displacement and 0.2Hz frequency, to simulate normal breathing at 12breaths/min. Table I shows the improvement of the output voltage and harvestable power for our miniature wearable electromagnetic sensor/harvester compared to piezoelectric modules.

Next the electromagnetic generator was tested on a healthy human subject. The module was placed on subject's chest, around sternum area. Fig.7 shows a sample output voltage (a), and power (b) of the sensor/harvester during normal breathing for a standing subject. Sensor was unobtrusive as reported by the subject, and respiratory cycles were easily observed. The maximum peak voltage was about 0.5 V, and maximum peak power several mW.

IV. CONCLUSIONS AND FUTURE WORKS

The selection criteria of a concurrent respiratory effort sensor and harvester, and optimization procedure, were presented in this paper. The comparison between piezoelectric materials and electromagnetic generators showed that the electromagnetic module is able to harvest more power, hundreds of µW to couple of mW, and detect respiratory effort. Future work will include improving the efficiency of the electromagnetic generator, and extracting physiological parameters from harvested signals.

V. ACKNOWLEDGMENT

The authors would like to thank Isar Mostafanezhad, Sangwoo Koo, and Noah Hafner for their assistance.

TABLE I. COMPARISON OF HARVESTED VOLTAGE & POWER BY MFC, PZT-5A, AND ELECTROMAGNETIC GENERATOR FOR A MECHANICAL TORSO

Measured Parameter	**Type of Sensor/Harvester Module**		
	M-8528-P2	***PZT-5A4E***	***Electromagnetic Generator***
Vmax (mV)	50	100.9	298.1
Pmax (µW)	15.6	63.63	555.39

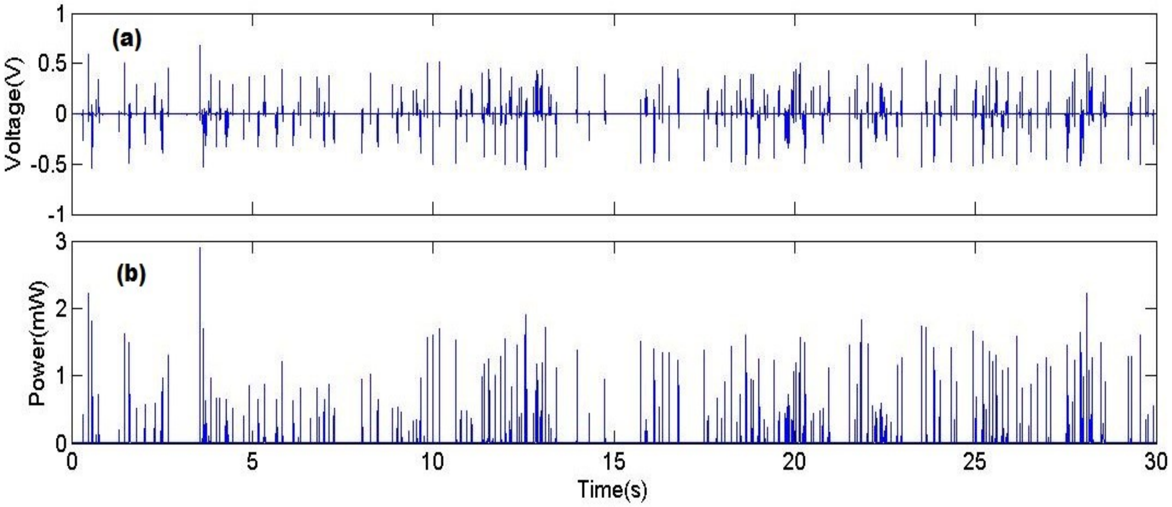

Figure 7. Output a) voltage, and b) power of the senror/harvester resulted from the movements of the chest during normal breathing for a male subject

REFERENCES

[1] T. Starner, “Human-Powered Wearable Computing” IBM Systems Journal, Vol.35, NOS 3&4, 1996.

[2] J. Paradiso and T. Starner, “Energy Scavenging for Mobile and Wireless Electronics,” Pervasive Computing, pp. 18-27, January-March 2005.

[3] T. Starner and J. Paradiso, “Human Generated Power for Mobile Electronics,” in Low Power Electronics Design, CRS Press, Fall 2004.

[4] J. Kymisis, C. Kendall, J. Paradiso, and N. Gershenfeld, “Parasitic Power Harvesting in Shoes”, Second International Conference on Wearable Computing, 1998.

[5] L. C. Rome, L. Flynn, E. M. Goldman, and T.D. Yoo, “Generating electricity while walking with loads,” Science, Vol. 309, Sept. 2005, Page(s): 1725-1728.

[6] R.Baghbani, M.Ashoorirad, “A Power Generating System for Mobile Electronic Devices Using Human Walking Motion”, Second International Conference on Computer and Electrical Engineering, 2009. ICCEE '09. Dec. 2009, pp: 385 – 388, Dubai.

[7] C.R. Saha, T. O’Donnell, N.Wang, P. McCloskey, “Electromagnetic generator for harvesting energy from human motion” Elsevier Sensors and Actuators A: Physical Vol. 147, Issue 1, 15 Sep. 2008, pp: 248-253.

[8] D.R. Goldhill, S.A. White, A. Sumner, “Physiological values and procedures in the 24 h before ICU admission from the ward”, Anaesthesia. 1999 Jun.;54(6):529–534.

[9] K.M., Hillman, P.J. Bristow, T. Chey, K. Daffurn, T. Jacques, S.L. Norman, et al. “Duration of life-threatening antecedents prior to intensive care admission”, Intensive Care Med. 2002 Nov.;28(11):1629–1634.

[10] J.F. Fieselmann, M.S. Hendryx, C.M. Helms, D.S. Wakefield, “Respiratory rate predicts cardiopulmonary arrest for internal medicine inpatients”, J Gen Intern Med. 1993 Jul.;8(7):354–360.

[11] M. Cretikos, J. Chen, K. Hillman, R.Bellomo, S. Finfer , A. Flabouris, et al., “The objective medical emergency team activation criteria: a case-control study”, Resuscitation. 2007 Apr.;73(1):62–72.

[12] J. Rosenberg, G.I. Rasmussen, K. R. Wøjdemann, L. T. Kirkeby, L. N. Jørgensen, H. Kehlet, “Ventilatory pattern and associated episodic hypoxaemia in the late postoperative period in the general surgical ward”, Anaesthesia. 1999 Apr.;54(4):323–328.

[13] T. Jonsson, H. Jonsdottir, A. D. Möller, L. Baldursdottir, “Nursing documentation prior to emergency admissions to the intensive care unit”, Nurs Crit Care. 2011 Jun.;16(4):164–169.

[14] G. G. Mazeika, and R. Swanson, “Respiratory Inductance Plethysmography An Introduction,” Pro-Tech Services, 2007: http://www.aastweb.org/resources/focusgroups/rip_intro.pdf

[15] O.Boric-Lubecke, V.Lubecke, I.Mostafanezhad, and E.Shahhaidar,”Zero-net energy wearable biosensors”, ISA 25th International Conf. on Computers and Their Applications (CATA2010), Honolulu,HI, Apr2010.

[16] E.Shahhaidar, O.Boric-Lubecke, M.Wolfe, and R.Ghorbani, ”Electromagnetic generators as respiratory effort energy harvester”, IEEE Power and Energy Conference (PECI), Urbana-Champaign, Illinoise, Feb.2011.

[17] B.Padasdao, and O.Boric-Lubecke, “Respiratory rate sensing using servo motors”, IEEE EMBC2011, Boston, MA, Aug.2011.

[18] E. Shahhaidar, E. Yavari, J. Young, O. Boric-Lubecke, C. Stickley,” Respiratory effort energy estimation using Doppler radar”, Annual International Conference of the IEEE Engineering in Medicine and Biology Society. IEEE Engineering in Medicine and Biology Society. Conference Aug. 2012.

[19] http://focus.ti.com/docs/prod/folders/print/msp430f2131.html

[20] Thomas P. Daue1, Jan Kunzmann, Andreas Schönecker, “Energy Harvesting Systems using Piezoelectric Macro Fiber Composites”, Smart Material Corp.

[21] http://www.me.mtu.edu/~wjendres/ProductRealization1Course/DC_Motor_Calculations.pdf